TITLE OF INVENTION

Continuous Tomography Bed Motion Data Processing Apparatus and Method

CROSS-REFERENCE TO RELATED APPLICATIONS

[0001] Not Applicable

STATEMENT REGARDING FEDERALLY SPONSORED RESEARCH OR DEVELOPMENT

[0002] Not Applicable

BACKGROUND OF THE INVENTION

1. Field of Invention

[0003] This invention pertains to apparatus and processes for three-dimensional image reconstruction from data acquired in a positron emission tomograph (PET). More particularly, this invention pertains to apparatus and methods based on a parallel/pipelined architecture for processing data acquired as the bed moves through the tomograph.

2. Description of the Related Art

[0004] In a positron emission tomograph (PET) imaging system, a patient is injected with a radioactively tagged substance that the body normally metabolizes in some fashion. The radioactive tag used is a positron-emitting isotope of either an element found in the substance or an element that is substituted for another element in the substance. For example, a widely used isotope is the positron-emitting isotope of fluorine, ^{18}F. This isotope is substituted, through a chemical synthesis process, for hydrogen in complex compounds such as glucose-forming fluro-deoxyglucose (FDG). When FDG is injected into a patient, the body will attempt to use it in the same fashion as it would normal glucose. Thus, there will be higher concentrations of positron emitters in areas where glucose is metabolized at higher levels, such as the brain, muscle tissue (the heart), and tumors.

[0005] As the FDG or other radiopharmaceutical isotopes decay in the body, they discharge positively charged particles called positrons. Upon discharge, the positrons encounter electrons, and both are annihilated. As a result of each annihilation event, gamma rays are generated in the form of a pair of diametrically opposed photons approximately 180 degrees (angular) apart. By detecting these annihilation "event pairs" for a period of time, the isotope distribution in a cross section of the body can be reconstructed. These events are mapped within the patient's body, thus allowing for the quantitative measurement of metabolic, biochemical, and functional activity in living tissue. More specifically, PET images (often in conjunction with an assumed physiologic model) are used to evaluate a variety of physiologic parameters such as glucose metabolic rate, cerebral blood flow, tissue viability, oxygen metabolism, and in vivo brain neuron activity.

[0006] Mechanically, a PET scanner consists of a bed or gurney and a gantry, which is typically mounted inside an enclosure with a tunnel through the center, through which the bed traverses. The patient, who has been treated with a radiopharmaceutical, lies on the bed, which is then inserted into the tunnel formed by the gantry. Traditionally, PET scanners are comprised of one or more fixed rings of detectors, surrounding the patient on all sides. Some newer scanners use a partial ring of detectors and the ring revolves around the tunnel. The gantry contains the detectors and a portion of the processing equipment. Signals from the gantry are fed into a computer system where the data is then processed to produce images.

[0007] Detectors on the detector rings encircling the patient detect the gamma rays, one on either side of the patient, and the time at which they were detected. Therefore, when two detectors on opposite sides of the patient have detected gamma rays that occurred within some time window of each other, it is safe to assume that the positron-electron interaction occurred somewhere along the line connecting the two detectors. If the detectors that detected the pair of gamma rays are located on the same ring, the coincidence plane, which is a transaxial plane, is called a direct plane. If the detectors are located on different rings, the coincidence plane, which is an oblique plane, is called a cross plane.

[0008] By histogramming the detected occurrences based on these lines of response (LOR), a pattern that uniquely describes the distribution of radioactivity is formed. The array in which the histogram is formed is typically called a sinogram. An image of the isotope distribution can be formed from these sinograms using any number of techniques that have been described in the prior art. However, the image that is produced is inaccurate due to several factors. As the gamma rays pass through the patient's body (and other objects, such as the patient bed), they are attenuated and scattered. Additionally, each gamma ray detector has a different response. All of these factors produce either noise or artifacts. Methods for correcting these effects are described in the prior art.

[0009] Positron emission tomography is one of the medical imaging modalities for which the transition from two-dimensional to three-dimensional acquisition has been most successful. Following pioneering work in the 1980s, the development after 1989 of multi-ring scanners equipped with retractable septa has led to the present widespread utilization of volume PET-scanners. These scanners have an open, collimator-less cylindrical geometry, which allows the measurement of coincidences between all pairs of detectors on the cylindrical surface.

[0010] Data collected in the transaxial or direct plane and in the oblique planes is three-dimensional (3D) data. These 3D data approximate line integrals of the radioactive tracer distribution along LORs which are not restricted to lie within transaxial planes. This is in contrast with the two-dimensional (2D) data acquired when the scanner is operated in 2D mode, in which the data collected is limited to LORs in the transaxial planes. The transition from 2D acquisition to 3D acquisition leads to a significant improvement of the scanner sensitivity, due to the increased number of measured LORs and to the elimination of the detector's shadowing by the septa.

[0011] Usually, 3D PET data are reconstructed using a reprojection algorithm (3DRP), which is a 3D filtered-backprojection (FBP) method obtained by discretizing an analytical reconstruction formula. Owing to the considerable number of LORs measured in 3D mode, it is not surprising that the 3DRP algorithm is much more time consuming than the 2D slice-by-slice FBP used to reconstruct data acquired in 2D mode. A further reason for this increased

complexity is that the reconstruction of the 3D image is not decomposed into the reconstruction of a set of independent slices. Other algorithms relying on exact analytical formulae have so far been unable to reduce reconstruction time by factors larger than 2 compared to the 3DRP algorithm. In contrast, significant improvements in the reconstruction speed have been achieved using various combinations of the three following approaches. The first one is the introduction of faster, but often expensive, hardware. The second approach uses a reduced sampling of the 3D data to decrease the number of LORs which must be backprojected. Reduced sampling is achieved by adding groups of adjacent LORs in such a way that the resulting loss of spatial resolution remains acceptable for a given type of study. Finally, the third approach to faster 3D reconstruction is the use of approximate algorithms based on axial rebinning. The Fourier rebinning (FORE) process is one such approximate algorithm. The FORE algorithm is described in "Exact and Approximate Rebinning Algorithms for 3D PET data," M. Defrise, P. Kinahan, D. Townsend, C. Michel, M. Sibomana, and D. Newport, IEEE Transactions on Medical Imaging, pp. 145-58, 1997.

[0012] The advantages of using a continuous axial scanning motion are described in "Implementation of True Continuous Whole Body PET Scanning," M. Dahlbom, J. Reed, and J. Young, IEEE 2000 Medical Imaging Conference. This paper describes performing a scan by moving the patient bed in small, discrete steps. True continuous movement of the patient bed is described in "Methods for Improving Image Quality in Whole Body PET Scanning," M. Dahlbom, DC Yu, S. Cherry, A. Chatziioannou, and E. Hoffman, IEEE Transactions on Nucl. Sci., Vol. 39, No. 4, pp. 1079-83, 1992. This second paper describes scanning a continuously moving subject and storing the data in list mode, which is later sorted into sinograms for reconstruction.

BRIEF SUMMARY OF THE INVENTION

[0013] Apparatus and methods for processing continuous bed motion, three-dimensional (3D) positron emission tomography (PET) acquisitions based on a parallel/pipelined architecture are provided. As the patient bed crosses predetermined positions, specific portions of the acquired data are inserted into the processing pipeline. At each stage of the pipeline, a different processing step is performed on the data in parallel to the others. One of these stages is the conversion of the 3D data set to a two-dimensional (2D) data set. The final result of the pipeline is a single reconstructed image plane corresponding to the acquired data inserted in the pipeline at an earlier time. As the patient bed moves, new image planes are continually produced in a periodic fashion. At the completion of the acquisition, only the portions of the data not in the pipeline and those remaining in the pipeline have to be processed through the pipeline.

[0014] During acquisition, the emission and/or transmission events are received from an acquisition processor, along with information on the current position of the patient bed. These events are histogrammed into a 3D sinogram space based on the current patient bed position. When the patient bed has moved a predetermined amount, the histogramming is shifted based on this amount. With this shift, a portion of the 3D sinogram space is no longer within the histogramming region, which corresponds to the portion of the patient and patient bed that has traversed, and is no longer within, the axial field-of-view of the tomograph. This portion of the 3D sinogram space is transferred to either an attenuation processing process (for transmission data) or a normalization process (for emission data). When normalization has been completed, the normalized emission data is transferred to an attenuation correction process. After attenuation correction has been completed, the corrected data is transferred to the Fourier Rebinning (FORE) process. The FORE process is a conversion of the data from a 3D data set to a 2D data set.

[0015] Just as with the histogramming process, when the patient bed has moved a predetermined amount, the FORE processing is shifted a corresponding amount. With this shift, a portion of the 3D sinogram space is no longer within the

FORE processing region. This region corresponds to the portion of the patient and patient bed that has traversed, and is no longer within, the axial field-of-view of the tomograph. This portion of the now 2D sinogram space is transferred to an image reconstruction process. After the reconstruction process is completed, the image plane is stored, scatter corrected, and/or displayed. All stages of this parallel/pipelined architecture are operating on data at the same time. However, the data for a given processing stage is different from the data in the other processing stages.

BRIEF DESCRIPTION OF THE SEVERAL VIEWS OF THE DRAWINGS

[0016] The above-mentioned features of the invention will become more clearly understood from the following detailed description of the invention read together with the drawings in which:

Figure 1 is a block diagram of the parallel/pipelined architecture;

Figure 2 is a perspective view of the detector rings of a scanner;

Figure 3 is a section view of the detector rings of a scanner;

Figure 4 shows the geometry of a cylindrical PET scanner;

Figure 5 shows the geometrical interpretation of Fourier rebinning;

Figure 6 is a Michelogram of the three-dimensional data acquired with a N=16 ring scanner;

Figure 7 is a series of Michelograms representing a series of bed positions;

Figure 8 is a pictorial illustration of the principle of a rebinning algorithm;

Figures 9A to 9D illustrate the representative lines of response for a four-ring scanner;

Figures 10A and 10B illustrate the acquired data of a source for a first bed position in a four-ring scanner;

Figures 11A and 11B illustrate the acquired data of a source for a second bed position in a four-ring scanner;

Figures 12A and 12B illustrate the acquired data of a source for a third bed position in a four-ring scanner;

Figures 13A and 13B illustrate the acquired data of a source for a forth bed position in a four-ring scanner; and

Figures 14A and 14B illustrate the acquired data of a source for a final bed position in a four-ring scanner.

DETAILED DESCRIPTION OF THE INVENTION

[0017] Apparatus and methods for processing continuous bed motion, three-dimensional (3D) positron emission tomography (PET) acquisitions based on a parallel/pipelined architecture are disclosed. A PET scanner has a bed that moves continuously as the patient is being scanned. The data from the scanner is processed as it is acquired, producing an image within a short time after the scanning is completed.

[0018] Figure 1 is a block diagram that illustrates the parallel/pipelined architecture. Although the following discussion is in terms of a process, the present invention includes the hardware and software used to implement the various process steps. The means to implement the individual processes are known in the art, as are the means to control the data flow between the processes. In one embodiment, the enumerated processes are implemented by a multi-threaded software program running on at least one processor. In another embodiment, a combination of hardware and software is used to implement the enumerated processes.

[0019] In Figure 1, the first block represents the acquisition **102** of the data from the scanners. The acquisition process **102** includes collecting the raw data from the scanner detectors and storing this data in a list mode data file. The data acquired includes emission and/or transmission events along with information on the current position of the patient bed. The acquisition process **102** collects data

continuously as the patient, on the patient bed, moves through the scanner. The data from the acquisition process **102** is output, asynchronously, to the histogram process **104**. The data stream to and from the acquisition process **102** is the only data stream that is asynchronous, all the other data streams to and from the other processes are synchronous, dependent upon equipment resources and the plane step time.

[0020] The histogram process **104** creates a 3D sinogram space histogram of the emission and/or transmission events received from the acquisition process **102**, along with information on the current position of the patient bed. Those skilled in the art will recognize that the bed position information can be either a time signal based on a fixed bed speed or a position signal based on a bed position sensor. The emission events are histogrammed into a 3D sinogram space based on the current patient bed position. When the patient bed has moved a predetermined amount, the histogramming is shifted a corresponding amount. With this shift, a portion of the 3D sinogram space is no longer within the histogramming region, which corresponds to the portion of the patient and patient bed that has traversed, and is no longer within, the axial field-of-view of the tomograph.

[0021] The histogram process **104** outputs synchronous data as two data streams **162, 156**. The first data stream **162** from the histogram process **104** transfers the contents of a transmission data file created during the histogram process **104** to a transmission/attenuation process **122**. The transmission data file contains two-dimension (2D) data. The transmission/attenuation process **122** uses an existing blank transmission data file to create an attenuation data file. The transmission/attenuation process **122** outputs a data stream to both an attenuation correction process **108** and a Mu image reconstruction process **124**. The Mu image reconstruction process **124** creates a Mu image data file and outputs a data stream to a scatter correction process **126**.

[0022] The second data stream **156** transfers the contents of a 3D emission data file created during the histogram process **104**. The second data stream **156** transfers the data to a normalization process **106**. The normalization process **106** uses an existing normalization file to create a second emission data file. The

existing normalization file contains the normalization coefficients. The normalization process **106** outputs a data stream to the attenuation correction process **108**.

[0023] The attenuation correction process **108** accepts a data stream from the transmission/attenuation process **122** and the normalization process **106**. The attenuation correction process **108** creates a sinogram data file and outputs a data stream to a Fourier rebinning (FORE) process **110**, which creates an image data file and outputs a 2D data stream to an image reconstruction process **112** and the scatter correction process **126**. The FORE process **110** converts the data from a 3D data set to a 2D data set.

[0024] The data passing through the FORE process **110** corresponds to the bed movement. After the patient bed has moved a predetermined amount, a portion of the 3D sinogram space is no longer within the FORE processing **110** region. This portion of the 3D sinogram space corresponds to the portion of the patient and patient bed that has traversed, and is no longer within, the axial field-of-view of the tomograph. The output of the FORE process **110**, which represents a 2D sinogram space, is transferred to an image reconstruction process **112**. After the reconstruction process **112** is completed, the image plane is stored, scatter corrected **126**, and/or displayed **114**.

[0025] The scatter correction process **126** accepts data streams from the image reconstruction process **112** and the Mu image reconstruction process **124**. The scatter correction process **126** creates a final image data file and outputs a data stream to the image display process **114**.

[0026] All stages of the above-described parallel/pipelined architecture are operating on data at the same time. However, the data for a given processing stage is different from the data in the other processing stages. Just as in any parallel/pipelined architecture, each stage of processing must complete processing the current data before accepting new data. Therefore, the data from one stage of processing cannot be sent to the next stage of processing until the next stage has completed processing data from the previous cycle. Thus, the overall speed of processing is determined by the slowest stage of processing. Those skilled in the

art will recognize that processing stages can be omitted or additional processing stages (various corrections, such as arc correction, etc.) can be added to the architecture without departing from the spirit and scope of the present invention.

[0027] Figure 2 illustrates a scanner **202** with 16 rings of detectors. Figure 3 is a cross-sectional view of the detector rings **302** in a scanner **202**. The indices **j** and **i** of Figure 3 each represent one half of a pair of rings in coincidence.

[0028] Figure 4 illustrates the geometry of a cylindrical PET scanner having a radius **R** and a length **L**. Figure 4 includes a transaxial view (left side of figure) showing the **x**, **y** axes and the sinogram variables **s** and ϕ, and it includes a longitudinal view (right side of figure) showing the **z** axis and the sinogram variables **z** and Δ. The axes **x**, **y** are shown rotated for illustrative purposes. In Figure 4, a line of response (LOR) is shown extending between two detectors **A** and **B**. The sinogram variable **s** is the distance between the **z** axis and the projection of the LOR onto a transaxial plane, and ϕ is the angle between this projection and the **y** axis, which, as illustrated in Figure 4, is equal to the angle of a line perpendicular to the projection of the LOR and the **x** axis. The set of data corresponding to a fixed pair (**s**, ϕ) define an ordinary, 2D sinogram.

[0029] The longitudinal view of Figure 4 shows the sinogram variable Δ as the longitudinal distance between the two detectors **A** and **B**, or $z_A - z_B$, with z_A being the location of one detector **A** along the **z** axis, measured from a point on the **z** axis, and z_B being the location of the other detector **B** along the **z** axis, measured from the same point on the **z** axis. A fourth sinogram variable, **z**, not illustrated, is defined as $(z_A - z_B) / 2$. Thus, **z** is the axial coordinate of the point mid-way between the two detectors, and Δ is the axial spacing between the two detectors **A** and **B**. The set of data corresponding to a fixed pair (**z**, Δ) define an oblique sinogram, with the special case of $\Delta = 0$ being called a direct sinogram. For a PET scanner with **N** rings, such as the 16 ring scanner illustrated in Figures 2 and 3, each pair of rings corresponds to a fixed pair (**z**, Δ), and hence, the data acquired in 3D mode consists of N^2 sinograms, in which are included **N** direct sinograms and $N \cdot (N-1)$ oblique sinograms. The four sinogram parameters (**s**, ϕ, **z**, Δ) define a 3D sinogram. A 2D sinogram is restricted to LORs in the transaxial plane, so that

$z_A = z_B$. Therefore, a 2D sinogram is defined by three parameters (s, ϕ, z). Reconstructing slices, or direct sinograms, from 2D data involves less parameters than reconstructing slices from 3D data. A rebinning algorithm is a method to estimate 2D slices from 3D data.

[0030] Figure 5 illustrates the geometry of a cylindrical PET scanner with a source **S** at a distance **t** from the axis **z**. The axial position of the source **S** can be determined from the equation $z' = z + t \bullet \tan(\theta) = z + \delta \bullet t$, where **z** is the axial point midway between the detectors **A** and **B** and where δ is the tangent of the angle θ between the LOR and the transaxial plane, called the ring difference. Although the distance **t** cannot be associated with an LOR, the 2D Fourier transform of the sinogram gives a frequency-distance relationship, which permits estimating the distance **t**. This relationship leads to the Fourier rebinning (FORE) algorithm. The FORE algorithm requires the 3D data to be organized as a set of oblique sinograms, each of which is characterized by a pair (z, δ). The algorithm processes each sinogram independently of the others, and its implementation is therefore independent of the way in which the variables **z** and δ are sampled.

[0031] Figure 6 illustrates a Michelogram **610**, which is a graphical representation of the planes of response which get grouped together to reduce data set size in three-dimensional PET. The Michelogram **610** illustrates the 3D data set acquired with a $N = 16$ ring scanner, as illustrated in Figures 2 and 3. The vertical **j** and horizontal **i** axes correspond respectively to the indices **j** and **i** of two rings in coincidence. The indices **j**, **i** are illustrated in Figure 3. Each square in Figure 6 corresponds to one oblique sinogram (i, j), which can be characterized by the pair (z, δ). The oblique sinograms are defined by the sampling scheme of the following equations:

$$\delta = d \bullet \Delta \bullet \delta \quad \text{where } d = (i - j) = 0, \pm 1, \pm 2, \ldots, \pm d_{max}$$

$$z = -(L - \sigma) / 2 + n \bullet \sigma/2 \quad \text{where } n = (i + j) = |d|, |d| + 2, |d| + 4, \ldots, 2N - 2 - |d|$$

where N = number of rings and the ring indices **j**, **i** run between 0 and N-1, $\sigma = L/N$, $\Delta \bullet \delta = \sigma/2R$ is the axial angular sampling and $-(L - \sigma) / 2$ is the axial

coordinate of the center of the first ring. The parameter d_{max} determines the maximum value of δ in the acquired data.

[0032] To gain both memory and reconstruction speed, 3D data is acquired with a reduced axial sampling as shown with the sampling scheme of the following equations:

$$\delta = \underline{d} \bullet \Delta \bullet \delta \quad \text{where } \underline{d} = 0, \pm S, \pm 2S, \pm 3S, \ldots, \pm d'_{max}$$

$$z = -(L - \sigma) / 2 + n \bullet \sigma/2 \quad \text{where } n = n_0, n_0 + 1, n_0 + 2, \ldots, 2N - 2 - n_0$$

where S is an integer parameter called 'span,' and $n_0 = \max\{0, |\underline{d}| - (S - 1)/2\}$. Each discrete sample ($\underline{d}$, n) is obtained by summing the range of LORs about $\underline{d}$ defined by $|d - \underline{d}| \leq (S - 1)/2$ where d is as defined above. The number of LORs summed in this manner is approximately S/2.

[0033] The reduced axial sampling scheme is illustrated in Figure 6, in which the sets of oblique sinograms linked by the diagonally oriented line segments are added together. The example shown is for a span S = 5 and d_{max} = 12. The central area **602** between the dashed lines represents where $\underline{d}$ = 0. Flanking that area is one area **604** that represents $\underline{d}$ = 5 and the another area **614** that represents $\underline{d}$ = -5. Extending outward from the center area **602**, the next area **606** represents $\underline{d}$ = 10 and the other area **616** represents $\underline{d}$ = -10.

[0034] Figure 7 is a 3D schematic of multiple groupings of 3D data as shown in Figure 6. As the patient bed traverses the scanner, a multitude of sinograms **702** are acquired. The data set **610** illustrated in Figure 6 represents the sinograms acquired and processed after the patient has traveled a specified distance, typically one-half the detector width. After the patient bed has traversed this distance, represented by the **k** axis in Figure 7, another set of sinograms is acquired and processed. The parallel planes representing the data sets of sinograms illustrate the data sets **702** generated during a PET scan with continuous bed motion. As the bed traverses the scanner, one data set **702** after another is acquired. Each of these data sets are processed independently and sequentially as illustrated in Figure 1.

[0035] For example, the acquisition process **102** continuously acquires raw data and outputs data to the histogram process **104**. When the histogram process **104** has processed a data set **702a**, it outputs that data set **702a** to the transmission/attenuation process **122** and/or the normalization process **106**, which processes the data and then outputs the data set **702a** to the next processing stage. Once the histogram process **104** outputs the data set **702a**, the histogram process **104** prepares to output the next data set **702b**, which can be output only when the transmission/attenuation process **122** and/or the normalization process **106** has completed its processing of the data set **702a** and has completed the transfer of the data set **702a** to the next stage. The data sets **702** flow through the parallel/pipelined architecture in this stepwise manner until all the data sets **702** acquired have been processed.

[0036] Figure 8 illustrates the principle of a rebinning algorithm and shows the basic steps in processing the data to produce a 3D image. Three-dimensional data **801** is acquired from the scanner and processed into N^2 oblique sinograms **804**, where N represents the number of direct slices or sinograms for the scanned image. The oblique sinograms are rebinned into 2N - 1 ordinary sinograms **806**, which represent slices separated by one-half the axial distance between adjacent detector rings. The rebinned data **806** is converted to 2N - 1 slices for the 3D image **808** by using a 2D FBP algorithm.

[0037] Figures 9A through 9C illustrate the coincidence planes for a four-ring scanner **902**. Figure 9D illustrates the placement of those coincidence planes on a Michelogram **904**, which is a graphical representation of the planes of response which get grouped together to reduce the data set size in 3D PET. Figure 9A shows the direct coincidence planes **0**, **2**, **4**, **6** and the pair of oblique coincidence planes **1**, **3**, **5**. Figure 9B shows the oblique coincidence planes **7**, **8**, **9** and Figure 9C shows the corresponding oblique coincidence planes **10**, **11**, **12**. Referring to the Michelogram **904** of Figure 9D, the coincidence planes **0** through **12** are indicated by the numbers in the cell corresponding to the coordinates of the rings **i**, **j** illustrated in Figures 9A through 9C.

[0038] As can be seen in Figures 9B and 9C, coincidence planes **8** and **11** define the maximum angle from the direct planes that an oblique plane will have

for the number of rings in the scanner **902**. This angle is called the acceptance angle. Figure 6 illustrates a Michelogram **610** for a 16-ring scanner **202** in which $d_{max} = 12$; that is, events detected by ring pairs more than 12 apart are not recorded. Accordingly, the acceptance angle used in Figure 6 is less than the maximum defined by the detector rings at opposite ends of the scanner **202**. The simplified example illustrated in Figures 9A through 9D does not illustrate reduced axial sampling as does the example illustrated in Figures 3 and 6.

[0039] The oblique coincidence planes **1**, **3**, **5** are cross planes and the events recorded in these planes are attributed to the space midway between the direct planes **0**, **2**, **4**, **6**. Because the cross planes **1**, **3**, **5** are defined by detectors in adjacent rings, the recorded events are summed. The oblique coincidence planes **7**, **9**, **10**, **12** are second-order cross planes with a plane separation of ± 2, and the events recorded in these planes approximately coincide with data recorded by the direct planes **0**, **2**, **4**, **6**. The oblique coincidence planes **8**, **11** are third-order cross planes with a plane separation of ± 3, and the events recorded in these planes approximately coincide with data recorded by the cross plane **3**.

[0040] The counting efficiency of the cross planes **1**, **3**, **5**, **7-12** is approximately twice that of the direct planes **0**, **2**, **4**, **6** because the cross planes **1**, **3**, **5**, **7-12** acquire data from twice as many detector pairs. To reorient the data acquired from the cross planes **1**, **3**, **5**, **7-12** into axial cross sections, the difference in counting efficiency must be corrected, which is done during the normalization process **106**.

[0041] Figures 10 through 14 provide a simplified illustration of the events recorded from a source **S** as it traverses a four-ring scanner **1012**. In Figure 10A, a scanner bed **1010** with a source **S** is positioned to traverse a four-ring scanner **1012**, which is shown in section. The scanner bed **1010** is shown with a source **S** at the midpoint of the first detector ring i_o, j_o. The only coincidence plane that can be detected at this point is illustrated as a dotted line extending from the upper portion of the first ring i_o, through **S**, and to the lower portion of the first ring j_o. Figure 10B illustrates a Michelogram showing a representation of the events recorded for the position of the source **S**. Diagonal to the Michelogram is a representation of the data set **1002**, **1004**, **1006** that is sent to the histogramming

process **104** after all the data is acquired, which occurs when the source **S** exits the scanner **1012**, as illustrated in Figure 14A. The acquisition data, or sinogram, **1004** is represented by the dot in the cell at the intersection of i_o and j_o and contains the acquisition data for the source **S** acquired by the detectors on the first ring i_o, j_o. This first acquired sinogram data **1004A** is stored in the data set storage point **1004**.

[0042] Figure 11A illustrates the source **S** after it has moved to the midpoint between the rings i_o, j_o and i_1, j_1. At this point, two coincidence planes are detected. In Figure 11B, the acquisition data for each coincidence plane is represented by the dots **1104A**, **1104B** in the Michelogram. The two dots **1104A**, **1104B** are connected by a solid line that indicates that the data from each coincidence plane is summed. The summed data **1104A**, **1104B** is added to the data stored in the data set storage point **1004**, which at this time includes only the sinogram data **1004A**.

[0043] Figure 12A illustrates the source **S** after it has moved to the midpoint of the second detector ring i_1, j_1. In Figure 12B, the three coincidence planes are represented by the three dots **1202A**, **1204A**, **1206A** in the Michelogram. The acquired data **1202A**, **1206A** is stored in the data set storage points **1002**, **1006**, respectively, and the data **1204A** for the vertical, or direct, coincidence plane is added to the contents of the data set storage point **1004**.

[0044] Figure 13A illustrates the source **S** after it has moved to the midpoint between the rings i_1, j_1 and i_2, j_2. In Figure 13B, two coincidence planes are represented by the two dots **1304A**, **1304B** in the Michelogram. The two dots **1304A**, **1304B** are connected by a solid line that indicates that the data from each coincidence plane is summed and added to the contents of the data set storage point **1004**.

[0045] Figure 14A illustrates the source **S** after it has passed through the scanner **1012**. The source **S** positions between those illustrated in Figure 13A and Figure 14A are not illustrated. At the point illustrated in Figure 14A, no more events can be recorded and the data set **1002**, **1004**, **1006** is transferred to the histogram process **104**. After the data set is transferred, the data set storage

locations **1002**, **1004**, **1006** are cleared and become available for storing another data set.

[0046] The example illustrated in Figures 10A through 14A traces a single point source **S** as it traverses a 4-ring scanner **1012**. Figures 2 and 3 illustrate a 16-ring scanner **202**, which has a more complex Michelogram **610**. Those skilled in the art will recognize that the number of rings in a scanner can vary without departing from the spirit and scope of the present invention.

[0047] From the foregoing description, it will be recognized by those skilled in the art that apparatus and methods for real-time three dimensional image reconstruction from data acquired in a positron emission tomograph (PET) has been provided. As the tomograph bed moves continuously through the scanner, the acquired data flows through a processing system with a parallel/pipelined architecture.

[0048] While the present invention has been illustrated by description of several embodiments and while the illustrative embodiments have been described in considerable detail, it is not the intention of the applicant to restrict or in any way limit the scope of the appended claims to such detail. Additional advantages and modifications will readily appear to those skilled in the art. The invention in its broader aspects is therefore not limited to the specific details, representative apparatus and methods, and illustrative examples shown and described. Accordingly, departures may be made from such details without departing from the spirit or scope of applicant's general inventive concept.